Chapter 21

INSTRUMENTATION FOR DOSIMETRY

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21.1. INTRODUCTION

Measurements of absorbed dose (or air kerma) are required in varying situations in diagnostic radiology. The radiation fields vary from plain, slit and even point projection geometry, and may be stationary or moving, including rotational. Owing to the use of low photon energies for these fields, it is important that dosimeters have a satisfactory energy response. In general, the requirements for dosimeter accuracy are less stringent than those in radiation therapy; however, the dose and dose rate measurements cover a large range.

Patient dosimetry (see Chapter 22) is a primary responsibility of the medical physicist specializing in diagnostic radiology and is required by legislation in many countries. Dose data are also required in the optimization of examinations for image quality and dose. Radiation measurement is also critical for occupational and public exposure control (see Chapter 24). Dose measurements are essential in acceptance testing and quality control (see Chapter 19).

Several types of dosimeter can be used, provided that they have a suitable energy response, but typically, ionization chambers of a few cubic centimetres in volume, or solid state detectors specifically designed for such measurements, are used. If dosimeters are used to make measurements during an examination, they must not interfere with the examination. These devices are also used for determination of the half value layer (HVL). Special types of ionization chamber are employed for computed tomography (CT), mammography and interventional radiology dosimetry.

21.2. RADIATION DETECTORS AND DOSIMETERS

21.2.1. General characteristics of radiation detectors

A dosimeter is an instrument that measures ionizing radiation. It usually comprises a measuring assembly, often referred to as an electrometer, and one or more detector assemblies, which may or may not be an integral part of the measuring assembly. In diagnostic radiology, dosimetric instruments can be classified as either active or passive dosimeters. Active dosimeters display the dose value directly and include ionization chambers and/or semiconductor detectors[1] (sometimes loosely referred to as solid state detectors) used to measure air kerma (K_i), air kerma rate ($\dot{K}_i$), air kerma–length product (P_{KL}) and air kerma–area product (P_{KA}) in the primary beam conditions.

Measurements involving scatter radiation, such as patient exit dose measurements and CT phantom measurements, are also performed with ionization chambers; however, care must be taken if attempting to use semiconductor detectors for this purpose. Passive dosimeters cannot display the dose value directly, but record a dose signal when exposed to radiation, which must be subsequently retrieved and converted to dose (or air kerma) by a reading device. These include solid state devices such as thermoluminescent dosimeters (TLDs), optically stimulated luminescent (OSL) dosimeters and film dosimeters (including radiochromic film) that may be placed on a patient's skin or inside cavities to measure the skin or organ doses. Similar measurements can be performed in phantoms.

Other instruments are needed to measure the X ray tube voltage and exposure time that can be used without direct connection into the electrical circuits of the X ray units. These non-invasive instruments are often called kV meters and timers. There are also a variety of devices used for occupational and public dose assessment, including ionization chambers for direct measurements and TLD, OSL and film for indirect use, as either personal dosimeters or area monitors (see Chapter 24).

21.2.2. Properties of diagnostic radiology dosimeters

Many types of diagnostic radiology dosimeter are commercially available for the measurement of air kerma (and its derivatives). They incorporate either ionization chambers or semiconductor detectors. Although ionization chambers have been the standard instruments for diagnostic radiology dosimetry for many years, semiconductor detectors have found widespread use recently in the area

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[1] Detectors need to conform to the IEC-61674 standard [21.1].

of quality control measurements, mainly because of their small size, ruggedness and convenience of use. The measurement assembly analyses and processes the electrical signals from the detector, in order to display the value of the radiological quantity being measured (K, $\dot{K}_i$, P_{KL}, P_{KA}) and its units, i.e. Gy, Gy/s, Gy·m or Gy·m^2, with SI subunit prefixes, e.g. m, μ or n. When an ionization chamber is used, the electrometer provides the appropriate polarizing voltage.

Most commercial dosimeters can be used for both radiographic and fluoroscopic applications, using either the accumulated air kerma over time (integrate mode) or air kerma rate mode. Some commercial dosimeters automatically perform conversion and/or corrections to their reading, in order to display the actual air kerma value. In most cases, the calibration coefficient is applied through the system's software, to convert the measured charge (current) to air kerma at a given beam quality. Some dosimeter models have internal sensors for measurement of the environmental temperature and pressure, in order to perform corrections for the air density automatically.

The air kerma, K (or any other associate dosimetric quantity), is obtained from:

$$K = M_Q k_{TP} N_{K,Q_0} k_Q \prod k_j \qquad (21.1)$$

where the dosimeter's reading M_Q is corrected for air density by k_{TP}, converted to air kerma at an appropriate reference radiation quality by the N_{K,Q_0} calibration coefficient and corrected for the applied X ray spectrum by the k_Q factor. Further corrections for other influencing quantities may be applied by multiplication factors k_j, for example, corrections for ion recombination, polarizing voltage, radiation incident angle or humidity (see Section 21.6).

Since the dosimeters are used for various types of X ray unit and exposure conditions, the choice of the appropriate instrument is important, in order for the radiation measurement to be sufficiently accurate. Irrespective of the application, radiation dosimeters must exhibit several desirable properties, as discussed below.

21.2.2.1. Sensitivity

Sensitivity is the minimum air kerma required to produce a signal output (charge or current produced by the detector and collected by the measuring assembly). The better the sensitivity of the dosimeter, the higher the charge (or current) produced for the same air kerma (rate) and consequently the better the air kerma (rate) resolution and detectability. Ionization chambers with larger active (effective) volumes exhibit higher sensitivity than those with smaller volumes.

For this reason, large ionization chambers are preferred for low air kerma rate measurements, such as in fluoroscopy or for scattered radiation. In radiography, where the air kerma rates are higher, smaller chambers can be used, allowing better measurement of spatial resolution.

In general, semiconductor detectors have a sensitivity that can be orders of magnitude higher than that of ionization chambers. This property, among others, makes the use of these detectors advantageous for a wide range of applications. However, their intrinsic energy dependence makes their use problematic in non-calibrated beams and for scatter radiation measurements.

21.2.2.2. Linearity

The dosimeter reading M should be linearly proportional to the air kerma (rate). All dosimeters exhibit a linear response for a certain range of air kerma (rate). The linearity range and the non-linear behaviour depend on the type of dosimeter and its physical characteristics. Among other factors, the scale/reading resolution of the measuring assembly, the sensitivity and the leakage/dark current of the dosimeter restrict the rated range to a lower value, while saturation (over ranging) effects determine the upper value. The air kerma (rate) range in which the dosimeter performance is linear (rated range) should be stated by the manufacturer; the linearity of the dosimeter over this range should be tested by the user.

According to IEC-61674 standard [21.1], the non-linearity of a dosimeter is expressed by the ratio $(R_{max} - R_{min})/(R_{max} + R_{min})$, which should be less than 0.02 over the whole rated range of air kerma (rate). Values R_{max} and R_{min} are the maximum and minimum dosimeter response, respectively, over the rated range of air kerma (rate). The response is the quotient of the indicated value (dosimeter reading) to the true value of air kerma (rate).

21.2.2.3. Energy dependence

For diagnostic dosimeters, the X ray spectrum (often referred to as the radiation or beam quality) is specified by the beam HVL and is one of the important quantities affecting the response of a dosimeter. Within the range of the clinical X ray radiation qualities (25–150 kV), the variation in the dosimeter response with energy may be significant. This depends on the detector type and its physical and structural properties. The variation in response to different radiation qualities is taken into account by the use of a beam quality correction factor k_Q (see Eq. (21.1)). For a radiation quality Q, k_Q is the ratio of the calibration factors for quality Q to the reference radiation quality (RQR 5, for example, see Section 21.6). By definition, k_Q is unity at the reference beam quality. Figure 21.1

shows the variation in k_Q with HVL for six commercial dosimeters, including both ionization chambers and semiconductor detectors. Simple semiconductor detectors generally have more pronounced variation of k_Q with energy; however, modern semiconductor detectors incorporate multiple semiconductor elements covered by filters (typically copper) that allow the necessary compensation to reduce the effect of radiation quality. The IEC-61674 standard [21.1] imposes a ±5% upper limit on variation of energy response in the 50–150 kV range, while the IAEA [21.2] proposes the stricter limit of ±2.6% for dosimeters used as reference instruments at calibration laboratories.
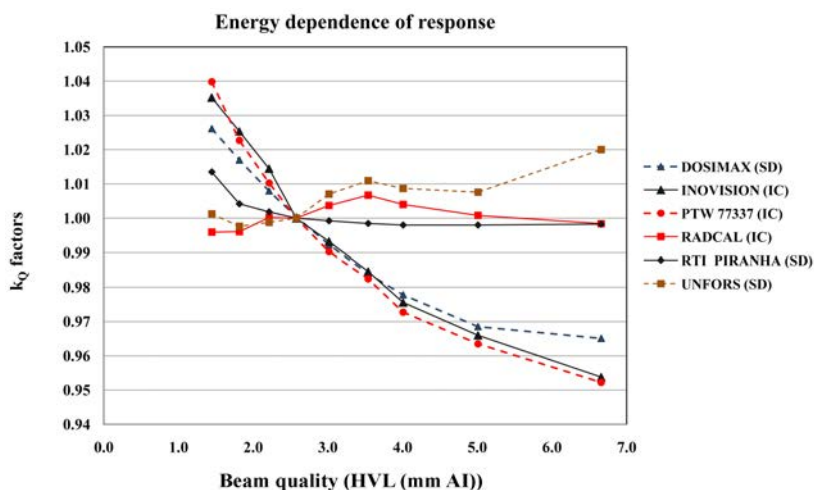

FIG. 21.1. Energy dependence of response of six commercial dosimeters incorporating ionization chambers (IC) or solid state detectors (SD): Dosimax plus (IBA, Schwarzenbruck, Sweden), Inovision Triad 35050A (FLUKE Biomedical Corp., Everett, WA, USA), PTW Unidos with 77337 (PTW GmbH, Freiburg, Germany), Radcal 2025 with 20x5-6 (Radcal Corporation Monrovia, CA, USA), RTI Piranha (RTI Electronics AB, Molndal, Sweden) and Unfors Xi (Unfors Instruments AB, Billdal, Sweden). The beam qualities (x axis) correspond to the RQR series described in the IEC-61674 standard [21.1].

21.2.2.4. Directional dependence

The response of a dosimeter may vary when the radiation is incident on the detector from different angles. The directional or angular dependence primarily depends on detector construction and physical size but will also depend on the energy of the incident radiation. The directional dependence of cylindrical or spherical ionization chambers is negligible, while parallel plate chambers might exhibit significant dependence at large incident angles. Most

commercial semiconductor detectors are mounted on lead backing plates, to attenuate radiation incident from the rear, while some models incorporate several semiconductor elements covered with filters to attenuate the radiation. In such cases, the directional dependence is important and care should always be taken to ensure that the radiation is incident on the elements through the filters at right angles. The IEC-61674 standard [21.1] imposes a ±3% upper limit of variation of response at incident angles of ±5° from the normal direction.

21.2.2.5. Leakage current

Leakage current refers to any signal change recorded by the measuring assembly that is not generated by radiation. This could be electronic noise, current from resistor–capacitor circuits, damaged cables or bad cable connections, lack of electronic or environmental equilibrium or humidity, etc. According to the IEC-61674 standard [21.1], the leakage current shall not exceed 5% of the minimum effective air kerma rate for the range in use. When a dosimeter is left in measurement mode after being exposed to the maximum effective air kerma value, the indicated value shall not change by more than 1% per minute.

21.3. IONIZATION CHAMBERS

The ionization detector is an air filled chamber in which an electric field is formed by the application of a polarizing voltage across two electrodes to collect all charges liberated by the ionization of the air contained within the chamber. The electric field is sufficient to collect almost all of the liberated charges that reach the electrodes (i.e. there is very little recombination) but insufficient to induce gas/charge multiplication and collision ionization of other molecules (in contrast with Geiger Müller and proportional counters). The number of ions collected, or the rate of their collection, is the recorded signal, which is multiplied by the mean energy required to produce an ion pair in dry air, $\bar{W}_{air} = 33.97 \, \text{eV} / \text{ion pair} = 33.97 \, \text{J/C}$ (see Eq. (3.12) and Sections 3.2.2 and 3.2.4 to deduce the energy transferred (ε_{tr}) from the radiation to the mass of air in the chamber). The ratio of ε_{tr} and the mass of air corresponds to the air kerma (rate) (Eq. (3.3)).

Figure 21.2 shows the internal structure of typical ionization chambers. In parallel plate chambers, the electrode separation is of the order of 1 cm and the electrodes are parallel to each other and to the entrance window. In cylindrical and spherical shaped chambers, the central electrode stands at the geometrical centre of the cavity, while the wall (outer shell) of the chamber is coated by a conductive material, which is often at ground potential (ground electrode). The wall (ground)

and the collecting electrode are separated by a high quality insulator to reduce the leakage current. A third electrode, the guard, reduces chamber leakage current by allowing any leakage to flow to ground, bypassing the collecting electrode and ensuring high uniformity of the electrical field in the chamber volume.
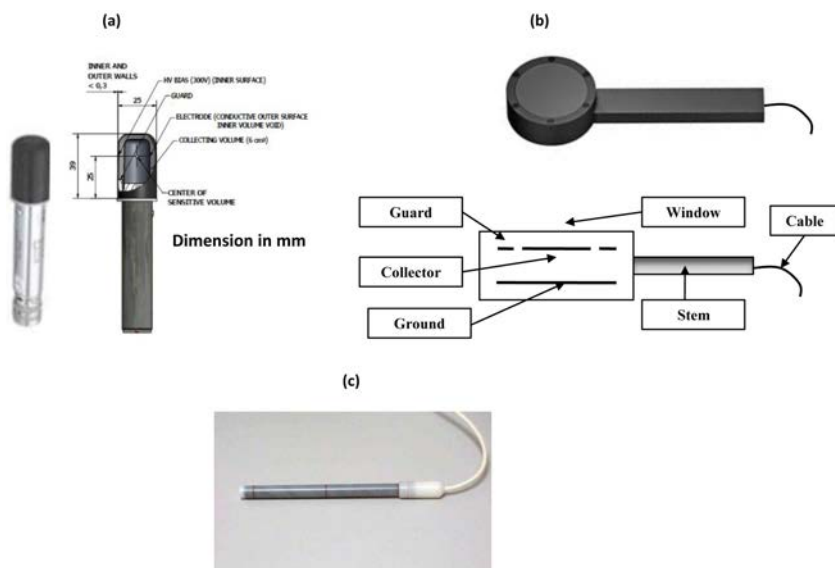

FIG. 21.2. Schematic diagram of (a) specialized chamber design (from http://www.radcal.com); (b) parallel plate (from http://www.standardimaging.com and Ref. [21.2]); and (c) cylindrical pencil type ionization chambers (from Ref. [21.2]).

Current high performance ionization chambers used in diagnostic radiology can have a more complex design, with the principle of keeping the gap between the ground and collecting electrodes small to prevent ion recombination at high dose being a primary consideration.

Ionization chambers used in diagnostic radiology should be vented, i.e. the air inside the volume communicates with the environment, rendering the mass of air dependent on temperature, pressure and humidity conditions. Humidity has an insignificant effect on air mass changes, but temperature and pressure affect the air mass within the chamber significantly. Therefore, the air density correction factor, k_{TP}, should always be applied to the dosimeter's readings. This factor is calculated from the formula $k_{TP} = (P_0 \cdot T)/(P \cdot T_0)$; where P_0 and T_0 are the values of the calibration reference conditions for pressure and temperature (usually 101.3 kPa (1 atm) and 293.2 K (273.2 + 20°C) or 295.2 K (273.2 + 22°C)), and P and T are the ambient pressure (kPa) and temperature (K) during the air kerma

measurement. According to the IEC-61674 standard [21.1], sealed chambers, in which the air volume does not change, are not suitable for diagnostic radiology dosimetry; their necessary wall thickness may cause unacceptable energy dependence, while the long term stability of the chambers is not guaranteed.

21.3.1. Clinical application of ionization chambers

21.3.1.1. Chambers for air kerma (dose) measurements

Determination of the air kerma (dose) in common diagnostic radiology applications (radiography, fluoroscopy and mammography) is performed by ionization chambers, either cylindrical or parallel plate. There is a large variety of chamber types and vendors.

Commercial parallel plate (p-p) chambers are disc shaped, with a diameter of several centimetres and a thickness of a few centimetres. The most common chambers with effective volumes (air cavity) from about 1 cm^3 to several hundreds of cubic centimetres are then suitable for application over a wide range of exposure rates. Owing to their shape, they can be safely inserted in hollow spaces, such as on the X ray table under a phantom, or in contact with the image intensifier, or inside the film cassette holder (Bucky), etc.

In mammography, p-p ionization chambers with a thin entrance window, made of a low density material (e.g. kapton film, acrylic, mylar) of micrometre thickness (20–50 μm, 3–10 mg/cm^2), are used. The major disadvantage of p-p chambers is the directional dependence of their response. The p-p chamber should always be placed perpendicular to the radiation beam.

Cylindrical chambers are uniformly sensitive around their central geometrical axis. The chambers used for measurement in the X ray beam have an effective volume of a few cubic centimetres (3–6 cm^3).

21.3.1.2. Cylindrical pencil type chambers

Cylindrical pencil type ionization chambers are used in several diagnostic radiology applications for the measurement of the air kerma length product, P_{KL}. For the last few decades, these chambers have mainly been used in CT dosimetry (see Section 22.4.7), but they are also used in dental applications (see Section 22.4.8). This chamber type is a long cylinder with a typical effective active length of 100 mm. The physical dimensions are about 15 cm in length and 1 cm in diameter (Fig. 21.2(c)). In contrast to other detectors used in diagnostic radiology, the chamber is partially irradiated. It is positioned with its axis at right angles to the central beam axis. The response of the active volume should be

uniform along its entire axial length. Special procedures and radiation qualities are used for their calibration.

21.3.1.3. KAP chambers

Air kerma area product (KAP) chambers have a large surface area and are transparent to both radiation and light. They are usually mounted on the tube housing after the beam collimation (see Sections 22.4.4 and 22.4.5) and encompass the entire radiation field. KAP chambers measure the integral of the air kerma over the area of the chamber and should have a uniform response throughout their entire area. Theoretically, P_{KA} is the same along the central X ray beam; however, in practice, scatter radiation, extra focal radiation and other factors affect the measurements. The requirement for the electrodes of the chamber to be transparent to light results in the use of materials that have a significant energy dependence over the diagnostic energy range.

Depending on their use and calibration, the KAP chambers measure the incident radiation, i.e. the radiation that falls on the chamber, or the transmitted radiation, i.e. the radiation that emerges from the chamber. The latter includes the attenuation of the radiation by the KAP chamber. Special procedures and radiation qualities are applied for the calibration of KAP meters.

KAP chambers are usually used for patient dosimetry in interventional radiology, fluoroscopy and general radiography, and are beginning to be used in pantomographic dental radiography (see Section 10.2.2.2). This is reflected in the use of KAP for diagnostic reference levels. Owing to the presence of extrafocal and scatter radiation, they should be calibrated in situ.

21.3.2. Application hints for ionization chambers

The practical points listed below should be considered:

- Appropriate ionization chambers should be selected for the application and the measuring procedure required (Table 21.1).
- Corrections for air density should always be applied to the dosimeter reading. Great care should be taken with dosimeters that incorporate internal sensors for automatic temperature and/or pressure corrections, in order to interpret the reading correctly.
- In general, ionization chambers detect radiations from all directions; thus, they measure all scatter, extrafocal and leakage radiation. When the incident air kerma is being measured, the chamber should be at a distance from the X ray couch or other supporting devices, in order to avoid backscatter

TABLE 21.1. BASIC CHARACTERISTICS OF DIAGNOSTIC RADIOLOGY DOSIMETERS

Application	Type of detector	Range of X ray tube voltage (kV)	Range of air kerma or air kerma rate	Intrinsic error (%)	Variation of energy response (%)	K rate dependence (%)	Angular dependence (%)
General radiography	Cylindrical, spherical or plane parallel IC[a] / Solid state detectors	60–150	10 µGy to 1 Gy / 1 mGy/s to 500 mGy/s[b] / 10 mGy/s to 5 mGy/s[b]	5	±5	±2	±3 @ ±5°
Fluoroscopy, interventional radiology[c]	Plane parallel IC[a] / Solid state detectors	50–120	10 µGy/s to 10 mGy/s[b,d] / 0.1 µGy/s to 100 µGy/s[b,d]	5	±5	±2	±3 @ ±5°
Fluoroscopy, interventional radiology[e]	KAP meters	50–150	10^{-1} to 10^6 µGy m^2 / 10^{-1} to 10^3 µGy m^2/s	10	±8	±5	—
Fluoroscopy	Plane parallel IC[a] / Solid state detectors / KAP meters	50–150	10 mGy/s to 10 mGy/s[a,b] / 0.1 mGy/s to 100 mGy/s[b]	5	±5	±2	±3 @ ±5°
Mammography	Plane parallel IC / Solid state detectors	22–40	10 µGy to 1 Gy / 10 µGy/s to 10 mGy/s[b]	5	±5	±2	±3 @ ±5°
CT	Cylindrical pencil type IC[a] of 100 mm active length[f]	100–150	0.1–50 mGy/s	5	±5	±2	±3 @ ±180°
Dental radiography	Cylindrical, spherical or plane parallel IC[a] / Solid state detectors / KAP meters / Cylindrical pencil type IC[a]	50–100	10 µGy to 100 mGy / 1–10 mGy/s	5	±5	±2	±3 @ ±5°

[a] IC: ionization chamber.
[b] Unattenuated beam.
[c] For air kerma rate measurements.
[d] In the light of new CT technologies and the revision of CT dosimetry methodology, new types of detector may be proposed that will be suitable for measuring pulsed radiation as well.
[e] For air kerma area product (rate) measurements.
[f] Attenuated beam.

radiation; other objects should not be allowed to interfere with the X ray beam.

- The ionization chamber should be totally covered by the radiation field, except for pencil type and KAP chambers. It is good practice to use field sizes at least twice the detector cross-section, and to check complete coverage of the detector through imaging methods if practicable. All chambers, especially p-p chambers, should be placed perpendicular to the radiation beam axis.
- Ionization chambers should be calibrated at several qualities. This is especially important for chambers with a large energy dependence. At least the qualities RQR 3 (50 kV), RQR 5 (70 kV) and RQR 9 (120 kV) should be used for radiography and fluoroscopy, and RQR-M1 (25 kV), RQR-M2 (28 kV) and RQR-M4 (35 kV) for mammography. For CT chambers, calibration should be performed at least at RQT 9 (120 kV) (see Section 21.6.2).
- The user should know the limitations and the rated ranges of all the quantities affecting the measurements. It is important to check that the leakage (dark) current is negligible and does not affect the measurements.
- Prior to use, users should check whether the battery voltage of the measuring assembly is within the manufacturer's rated range.

21.4. SEMICONDUCTOR DOSIMETERS

Diagnostic radiology dosimeters based on semiconductor technology have found widespread use. Two types are used: silicon diodes or metal oxide semiconductor field effect transistors (MOSFETs). Owing to their small size and rigidity, they are convenient for use in many applications.
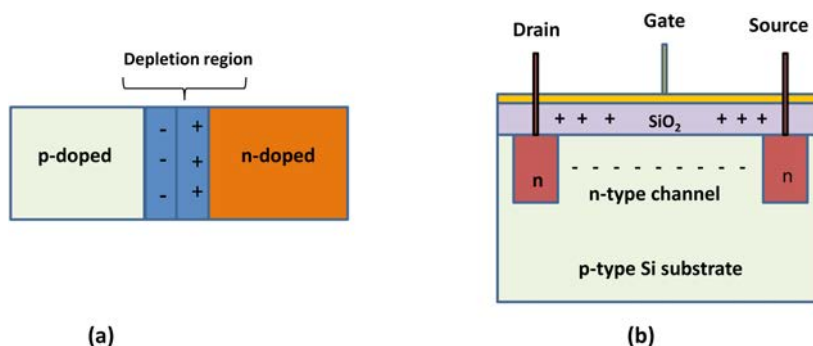

FIG. 21.3. Cross-sectional diagram of (a) p–n junction and (b) MOSFET.

21.4.1. Theory of operation

A silicon diode dosimeter is a p–n junction diode. In most cases, p type (rather than n type) diodes are used for diagnostic radiology dosimeters, since they are less affected by radiation damage and have a much smaller dark current (noise). When radiation falls on the diode, it produces electron hole pairs in the body of the diode and a current is generated in the reverse direction in the diode. The number of such pairs is proportional to the incident radiation dose. Owing to the diode structure and the intrinsically formed potential difference, there is no need to apply a bias voltage across the p and n type diode regions to collect the charge liberated by the radiation.

A MOSFET is a miniature silicon transistor. Its structure is equivalent to a planar capacitor with one of the electrodes replaced by a semiconductor. MOSFET dosimeters are based on the production of electron hole pairs in the SiO_2 of the MOSFET gate region (Fig. 21.3) resulting from incident radiation. The positive charge carriers move in the direction of the Si–SiO_2 interface, where they are trapped, building up a positive charge, which causes changes to the current in the n type channel and leads to a change of the gate bias voltage (shift in the threshold voltage). The threshold voltage shift is a linear function of absorbed dose. The integrated dose may be measured during (in real time) or after irradiation. MOSFETs often require a connection to a bias voltage during irradiation. They are mainly used in patient dosimetry.

21.4.2. Application hints for semiconductors

The practical points listed below should be considered:

- The response of semiconductors (diodes and MOSFETs) generally has a more pronounced energy dependence than that of ionization chambers. Although modern dosimeters typically use compensation methods to correct the energy dependence at specified beam qualities, the energy dependence for non-specified beam characteristics may be unpredictable. The user should investigate the dosimeter's energy dependence characteristics. In this respect, measurements of the HVL with semiconductor detectors should be avoided.
- The angular dependence of semiconductor detectors is comparable to plane parallel ionization chambers. However, semiconductor detectors are sensitive to their positioning in the X ray field, especially to the direction of the heel effect.
- When a semiconductor detector is used for dose measurements on a surface of a phantom (or patient), backscatter and sidescatter radiation may not

contribute significantly to the dosimeter reading, owing to the presence of backing plates.

- The semiconductor detector response does not depend on temperature or pressure. For the sake of a standard dosimetric formalism, k_{TP} in Eq. (21.1) is set to unity.

- Semiconductors have a limited useful life, owing to accumulated radiation damage. Although the doses measured in diagnostic radiology dosimetry are low, it is good practice to recalibrate the detectors at regular intervals.

- Research with MOSFET devices is currently in the experimental stages for dose measurements in some aspects of diagnostic radiology. These may be potentially beneficial in some high dose applications, such as interventional radiology, where high skin doses need to be avoided. However, they exhibit a high energy dependence and, therefore, frequent calibration is essential in order to achieve adequate measurement accuracy.

21.5. OTHER DOSIMETERS

21.5.1. Film dosimetry: Radiographic film and radiochromic film

21.5.1.1. Radiographic film

Radiographic film still finds application as a dosimeter in personal radiation monitoring using film badges (see Section 24.5.3). The structure of radiographic film and the basic principles of densitometry are described in Section 7.3.3. The emulsion in a film dosimeter directly absorbs ionizing radiation and can be correlated to the optical density of the developed film. However, the sensitometric curve is very different to that for screen film systems. A radiographic emulsion is far from tissue equivalent and the energy response of a film badge is, therefore, modified by addition of several filters. The provision, processing and analysis of such dosimeters are the tasks of specialized departments and companies and are not commonly within the duties of a medical physicist.

21.5.1.2. Radiochromic film

Radiochromic films (e.g. Gafchromic®) contain colourless dyes (diacetylene) that become blue after exposure because of radiation induced polymerization. This process is self-developing and requires no chemical process but needs some time for full development. Depending on the material, a density increase of about 10% from 1 to 24 h after exposure is typical.

The film comprises an active dye layer (15–20 µm thick) sandwiched between two transparent polyester sheets, each containing a yellow dye. The yellow dye enhances visual contrast and reduces the effects of exposure to blue and ultraviolet light. Some films use an opaque white backing sheet. Film optical density is measured with densitometers or film scanners. For films with an opaque backing, a reflective densitometer is needed. The blue coloured polymer exhibits a maximum in optical absorption at around 635 nm. Accordingly, a densitometer with a red light source should be used.

The composition of the film is near tissue equivalence. Some types of film incorporate barium compounds in the white backing to increase radiation absorption and sensitivity. Several types of radiochromic film are optimized for applications in diagnostic radiology. Their energy response and other properties can differ and the specifications should be collected from the supplier or from the literature. Sensitivity ranges from ~1 mGy to ~50 Gy, depending on film type. The sensitometric response is not linear and suitable calibration curves need to be applied. For film calibration and dose measurements, it is essential to use the same protocol and densitometer. The handling of radiochromic films is simple. Darkrooms are not required and ambient conditions are of little concern except for exposure to intensive light sources or humidity. The film can be obtained in large format (35 cm × 43 cm, maximum) and can be bent and cut to size as required.

Radiochromic films can be used for relative dosimetry in diagnostic radiology. The measurement and mapping of patient skin dose in interventional procedures is one such application (see Chapter 8).

21.5.2. Thermoluminescent dosimetry

A large and growing number of solid state materials exhibit the phenomenon of thermoluminescence (TL), which can be harvested for dosimetric purposes. This process consists of two stages: the first stage is the transference of an equilibrium TLD material to a metastable state through irradiation, and the second stage is application of energy (through heat) to reduce the metastable state back to equilibrium. Figure 21.4 demonstrates these two stages, using a semiconduction model for the solid state material. In this model, electron energies are not localized and the narrow energy gap between the valency and conduction bands is populated with midgap or trap sites that are caused by defects within the material.[2] Irradiation creates free electrons with enough energy to cross the gap into the conduction band, with the possibility of some of the electrons being

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[2] Trap sites are usually formed by the addition of dopant material. For example, TLD100 is made of LiF with additional dopant materials of Mg and Ti.

trapped in a midgap site. By subsequently adding energy to the material, trapped electrons with sufficient energy can escape the trap site into the conduction band and might return towards the valency band to recombine with a trapped hole, accompanied by a radiative TL emission (see Chapter 7 for a discussion on photostimulable phosphors).
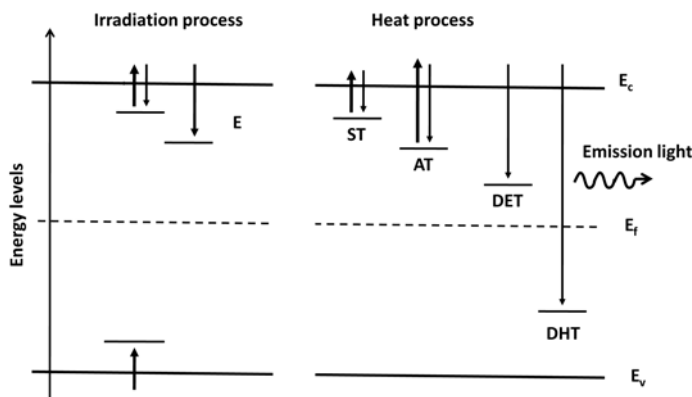
FIG. 21.4. Electron energy levels in a TLD material showing on the left the process of free electron and hole creation, followed by non-radiative charge trapping. On the right, the release of thermally stimulated electrons is shown for energy level $E_c - E$. The released electron may be retrapped or recombine with trapped holes. If this process is radiative, TL emission occurs. E_c and E_v are the conduction and valency band edges, E_f is the Fermi level, ST and AT are shallow and active traps, respectively, while DET and DHT are deep electron traps and deep hole traps, respectively.

The stability of the trapped electrons depends largely on the energy level (depth) of the traps. Shallow traps require little energy for electron release and are thermally unstable, leading to signal fading at ambient temperatures. Trap energy levels corresponding to higher excitation temperatures are desirable to obtain stable signals.

In a typical TLD reader (Fig. 21.5), the dosimeters are placed on a planchet heated directly by an electric current. The temperature is measured with a thermocouple welded to the planchet. Other methods of heating the TLD are also used, such as hot nitrogen jets, laser heating or infrared lamps. The TL signal is detected with a photomultiplier tube.

If a linear temperature ramp is applied, the TL signal (glow curve) shows various peaks at characteristic temperatures attributable to the traps present. Figure 21.6 shows a typical glow curve for LiF:Mg,Cu,P. Besides peaks at lower temperatures, the main peak useful for dosimetric measurements appears at ~210°C. Each type of TLD requires a specific optimized reading cycle.
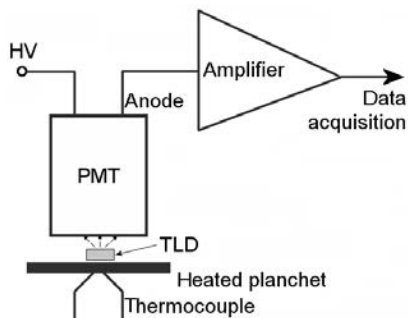
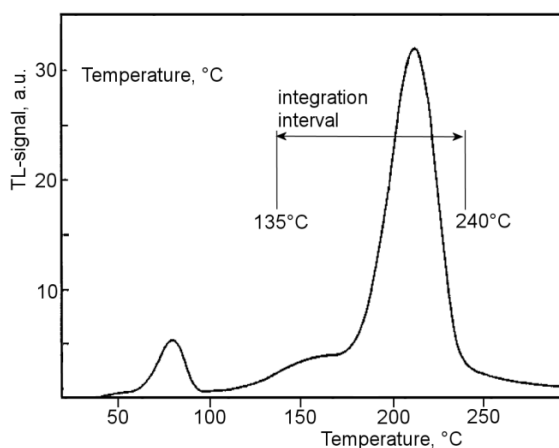FIG. 21.5. Principal elements of a TLD reader (PMT: photomultiplier tube).

FIG. 21.6. Typical glow curve for LiF:Mg,Cu,P (a.u.: arbitrary units).

The reading cycle of a TLD is divided into preheat, signal integration and annealing. During preheat, the dosimeter is maintained for some seconds at a constant temperature sufficient to remove all low temperature signals. The temperature is then raised up to the maximum value. In that period, the TL signal from the dosimeter is integrated to give a dose relevant signal. A typical integration temperature interval is shown in Fig. 21.6. Finally, the dosimeter is annealed in a dedicated oven to remove all remaining signals, thus resetting the dosimeter to zero. The reading cycle parameters depend on the TLD material, and good reproducibility is essential to achieve accurate results using the same reader for calibration and measurements.

The commonly used LiF:Mg,Ti (e.g. TLD100) is a well standardized dosimeter but less sensitive than LiF:Mg,Cu,P (GR200, TLD100H, MCP-N), which has a detection threshold of about 0.1 μGy. TLDs are available in many

forms and shapes (chips, rods, cubes, ribbons and powder). The relationship of dose to TL signal is linear up to doses of <1 Gy. For higher doses, correction factors for a non-linear response can be applied.

21.5.3. OSL

OSL is the luminescence emitted from an irradiated solid state material (OSL dosimeter) after being illuminated by stimulating light. The absorbed radiation causes ionization of the valence electrons and creation of electron pairs and leads the material to a metastable state. Pre-existing defects within the material localize the free electrons and holes through non-radiative trapping transitions. The illumination of the irradiated material with stimulating light (visible or infrared) leads to electron transition from the localized trap into the conduction band (equilibrium, ground energy level) and subsequent radiative emission and luminescence (OSL). In terms of energy levels, the two processes (excitation and transition) are quite similar to those described in Fig. 21.4. In fact, OSL is closely related to TL, with the basic difference being the use of light instead of heat as the added energy for the trapped electron. OSL should not be confused with photoluminescence, where the electron excitation is caused by light absorption rather than radiation dose.

Usually, the stimulating light used for OSL has a lower photon energy than the emitted light. The intensity of the emitted light is related to the rate at which the system returns to equilibrium, resulting in a characteristic luminescence–time curve. The integral of this curve corresponds to the trapped charged concentration, which is proportional to the radiation absorbed dose.

In a typical measurement using an OSL dosimeter, the sample material is illuminated with an appropriate light source. The emitted light is passed through an optical filter to suppress unwanted (PL) light and then detected with a photomultiplier tube. The arrangement of such a reader is similar to a TLD reader. An improvement in signal to noise ratio can be achieved by pulsing the stimulating light.

One OSL dosimeter that is commercially available uses aluminium oxide doped with carbon (Al_2O_3:C) and its dominant OSL trapping levels require thermal energies above 200°C to create thermoluminescence. Consequently, the OSL signal is thermally stable and signal fading is negligible. Some transient signals due to shallow traps will disappear after a few minutes. Dominant emission occurs in a band centred at around 420 nm. Stimulation of OSL is carried out by green light, from either green light emitting diodes or a laser. Since a single reading with useful signal intensities requires only 0.05% of the signal, stored re-reading or intermittent reading of the dosimeter is feasible and the dosimeter can be kept as a permanent dose record. Care must be taken to avoid

exposure of the dosimeter to light (particularly ultraviolet), as electrons from deep traps could be transferred to dosimeter traps (phototransfer), changing the response of the dosimeter. Doses in the range 10 µGy to 15 Gy can be measured using commercial systems. The OSL principle is also utilized for imaging using computed radiography[3] systems (see Chapter 7).

21.5.4. Dosimetric applications of TLD and OSL

Solid state dosimeters can be used for patient dosimetry external to the body or phantom in the same way as an ionization chamber, for internal measurements, typically in a phantom and also for occupational and public exposure monitoring (Section 24.5.3).

For internal dosimetry, a near tissue equivalent composition may have advantages in determining energy deposition within the body (see Section 2.4.2). The effective atomic numbers of tissue and water are 7.22 and 7.42, respectively. TLD materials such as LiF:Mg,Ti and $Li_2B_4O_7$:Mn have effective atomic numbers of 8.31 and 7.4, respectively, while the main OSL material, Al_2O_3, and the promising OSL material, BeO, have effective atomic numbers of 11.3 and 7.21, respectively.

It must be remembered, however, that the primary dosimetry system in diagnostic radiology is based on air kerma and not absorbed dose to water or tissue. Further, solid state dosimetry is a relative methodology that requires standardized calibration procedures. Care must be exercised if TLD or OSL dosimeters are used in radiation fields that differ from the calibration conditions. Consequently, careful consideration must be given before LiF and Al_2O_3 dosimeters are used in applications such as CT phantom dosimetry.

21.6. DOSIMETER CALIBRATION

All instruments used for dosimetric measurement in the clinical environment should have a calibration traceable to a recognized dosimetry standard. A prerequisite for measurement of a dosimetric quantity, such as air kerma, is that there be an international measurement system that determines the quantity and its unit. Primary standards dosimetry laboratories (PSDLs) employ free air ionization chambers for the measurement of absorbed dose traceable to the fundamental SI absorbed dose unit (Gy). The secondary standards dosimetry laboratories (SSDLs) calibrate their reference class instruments at PSDLs and

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[3] In the case of computed radiography, the active imaging material is known as a photon stimulable phosphor.

use these as their local dosimetry standards. Therefore, the traceability of the measurements to the specific PSDL is maintained. The main role of the SSDL is to bridge the gap between a PSDL and the dosimeter user.

21.6.1. Standard free air ionization chamber

Free air ionization chambers are often used by PSDLs as the primary standard for the determination of air kerma against which the secondary standard chambers from SSDLs are calibrated. Such a chamber is shown in Fig. 21.7. The charge liberated by X rays in the mass of the air inside the chamber volume is measured. The air kerma is deduced according to its definition, $K = (dE_{tr})/dm = (dQ \cdot W_{air})/dm$, from measurements of basic physical quantities (charge and mass) and applying physical constants and relative correction factors (see Chapter 3).

21.6.2. SSDL calibration

Most SSDLs apply the substitution method for the dosimeter calibration. At a given beam quality, Q, the true value of air kerma, $K_Q^{\text{'true'}}$ is measured using the reference dosimeter. The reference point of the user's dosimeter is placed at the same point and the dosimeter's reading is used to derive the calibration coefficient from the ratio $N_{K,Q}^{\text{user}} = K_Q^{\text{'true'}}/M_Q^{\text{user}}$, where M_Q^{user} is the reading of the user's instruments corrected for air density.
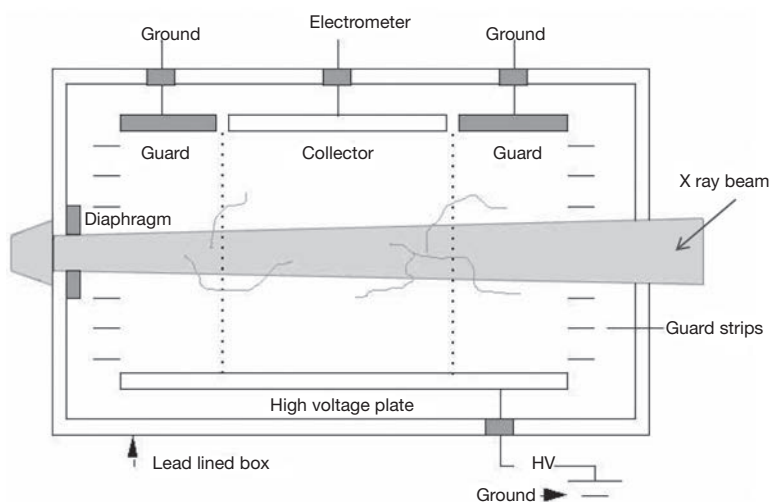

FIG. 21.7. The standard free in air ionization chamber used for calibration of an ionization chamber dosimeter.

The calibration of diagnostic radiology dosimeters is performed according to the radiation qualities that are described in the IEC-61267 standard [21.3] and is achieved using appropriate tube filtration at the specified tube voltage. Depending on the application of the dosimeter, a series of different beam qualities is used. For example, the RQR series simulates the primary beams incident on the patient, the RQT series simulates the beam qualities used in CT, the RQA series simulates the transmitted radiation qualities through the patient, and the RQR-M series simulates mammography beams. Each series consists of several beams with different combinations of tube voltage and filtration (see Tables 21.2–21.4). For the RQR, RQA and RQT quality series, an X ray tube with a tungsten (W) target and aluminium (Al) and/or copper (Cu) filters is used, while for the RQR-M quality series, an X ray tube with a molybdenum (Mo) target and molybdenum (Mo) filter is used.

TABLE 21.2. CHARACTERIZATION OF RADIATION QUALITY SERIES RQR USED FOR UNATTENUATED BEAMS FOR GENERAL RADIOGRAPHY APPLICATIONS
(according to IEC-61267[21.3])

Radiation quality	X ray tube voltage (kV)	First HVL (mm Al)	Homogeneity coefficient (h)
RQR 2	40	1.42	0.81
RQR 3	50	1.78	0.76
RQR 4	60	2.19	0.74
RQR 5[a]	70	2.58	0.71
RQR 6	80	3.01	0.69
RQR 7	90	3.48	0.68
RQR 8	100	3.97	0.68
RQR 9	120	5.00	0.68
RQR 10	150	6.57	0.72

[a] This quality is generally selected as the reference of the RQR series.

TABLE 21.3. CHARACTERIZATION OF RADIATION QUALITY SERIES RQR-M USED FOR UNATTENUATED BEAMS FOR MAMMOGRAPHY APPLICATIONS
(according to IEC-61267[21.3])

Radiation quality	X ray tube voltage (kV)	First HVL (mm Al)
RQR-M1	25	0.28
RQR-M2[a]	28	0.31
RQR-M3	30	0.33
RQR-M4	35	0.36

[a] This quality is generally selected as the reference of the RQR-M series.

TABLE 21.4. CHARACTERIZATION OF RADIATION QUALITY SERIES RQT USED FOR CT APPLICATIONS
(according to IEC-61267[21.3])

Radiation quality	X ray tube voltage (kV)	First HVL (mm Al)
RQT 8	100	6.90
RQT 9[a]	120	8.40
RQT 10	150	10.1

[a] This quality is generally selected as the reference of the RQT series.

A general purpose dosimeter should be calibrated in terms of air kerma at the RQR (RQR 2 to RQR 10) radiation qualities. According to common practice, the calibration coefficient, N_K, of a dosimeter is obtained at the RQR 5 (70 kV). For the other radiation qualities of the RQR series, further correction factors (k_Q) are provided to take into account the energy dependence of the dosimeter response. For a given radiation quality Q, k_Q is defined as the ratio of the calibration coefficients at radiation quality Q to that at radiation quality RQR 5. By definition, $k_Q = 1$ at RQR 5. For mammography, the standard beam quality is RQR-M2 (28 kV) and for CT it is RQT 9 (120 kV).

21.6.3. Field calibration

In some cases, for practical, economic and other reasons, users may calibrate their field instruments themselves. For example, when many dosimeters are being used in a large hospital, the user may prefer to calibrate them against a reference dosimeter, rather than send all of them to an SSDL. Some dosimetry equipment, such as KAP meters, is permanently installed on X ray systems and

must be calibrated on-site. Generally, cross-calibration of a field instrument refers to its direct comparison in a suitable user's beam quality, Q, against a reference instrument that has been calibrated at an SSDL. The calibration coefficient is obtained from Eq. (21.2):

$$N_{K,Q}^{\text{field}} = \frac{K_{i,Q}^{\text{ref}}}{M_Q^{\text{field}}} = \frac{M_Q^{\text{ref}} N_{K,Q_0}^{\text{ref}} k_Q^{\text{ref}}}{M_Q^{\text{field}}} \tag{21.2}$$

where 'field' and 'ref' refer to the field and the reference instruments, respectively. The M values are readings of the reference and the field instruments and have been corrected for the influence of all quantities except beam quality. Since the calibration coefficient refers to a specific beam quality, the cross-calibration should be performed at the whole range of beam qualities that are used in the hospital. It is important to note that other essential elements of traceability of measurement, such as uncertainty evaluation, evidence of competence, documentation, etc., should be taken into account and be declared for cross-calibrations.

21.7. INSTRUMENTS FOR MEASURING TUBE VOLTAGE AND TIME

Measurement of the X ray tube voltage and exposure duration (often referred to as 'exposure time') is usually performed with non-invasive, portable electronic devices, often called kV meters and timers.

Figure 21.8 shows a typical X ray tube voltage waveform from a three-phase, six-pulse generator operating at an 80 kV tube voltage and a 165 ms exposure time. Depending on the model, the kV meter measures the absolute peak voltage (the maximum value of the voltage during the exposure — circled point in Fig. 21.8), the average peak voltage (average of all peak values), the average voltage (average of all voltage values), the effective peak voltage (the voltage that would give the same image contrast as a constant potential X ray system) and the practical peak voltage (defined as the equivalent value of a voltage of any waveform related to an ideal X ray generator that provides a constant voltage and that produces the same image contrast). Practical peak voltage has been proposed as the standard quantity for the X ray tube voltage.

The kV meter is positioned in the primary X ray beam and measures the X ray tube voltage with methods based on attenuation measurements. Such instruments usually incorporate two (or more) detectors covered with filters (usually made of copper) of different thickness. When exposed to radiation, the detectors produce different signals, owing to the different attenuation of the X ray

beam by the filters. The signal ratio (or any other relationship of the signals) is a function of the incident X ray energy and consequently of the tube voltage. During the initial calibration of the kV meter at the factory, the signal output and/or the reading is appropriately adjusted to the 'correct' tube voltage value. Many kV meters digitize, process and store their detector signals and can supply voltage and/or exposure waveforms. The kV meter detectors' long geometrical axis should be positioned perpendicular to the tube anode–cathode direction, to eliminate the influence of the 'heel' effect to the kV measurement.
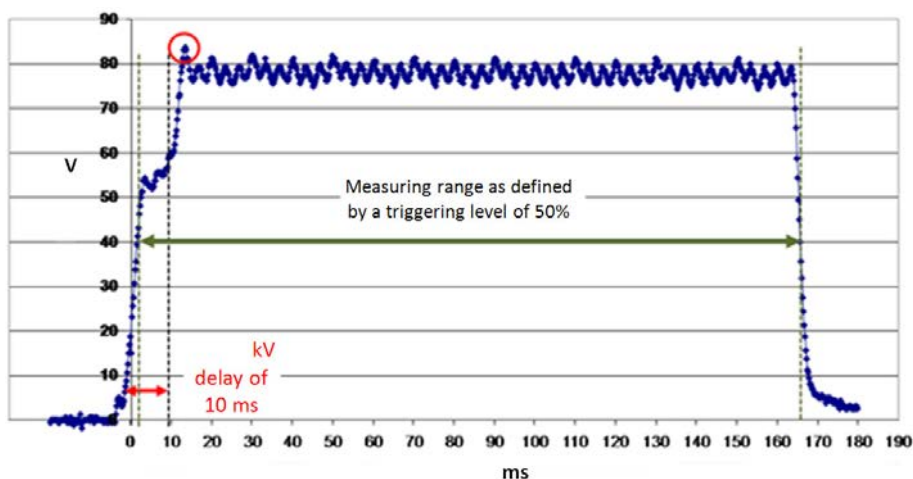

FIG. 21.8. Typical X ray tube voltage waveform from a three-phase six pulse generator operating at 80 kV tube voltage and 165 ms exposure time.

The IEC-61676 standard [21.4] specifies the performance requirements of instruments used for the non-invasive measurement of the X ray tube voltage up to 150 kV. It recommends that the relative intrinsic error of the practical peak voltage measurement should not be greater than ±2% over the effective voltage ranges, i.e. 60–120 kV for diagnostic and 24–35 kV for mammography. In addition, it recommends that a 1.5% limit of variation in response is acceptable at tube filtration ranges of 2.5–3.5 mm Al (for diagnostic radiology applications).

Exposure time is the time during which the radiation X ray beam is produced. It is measured as the radiation pulse width (time difference) between an 'initial' and 'final' point of the exposure, which are defined by a preset triggering level. The proposed convention for timing measurements of X ray systems is to measure the pulse width at a height of 50% of the waveform peak (full width at half maximum). Some manufacturers use different values of the triggering level

(e.g. 10% or 75%). The exposure time may be measured using either invasive or non-invasive equipment.

21.8. INSTRUMENTS FOR OCCUPATIONAL AND PUBLIC EXPOSURE MEASUREMENTS

Radiation monitoring is performed in diagnostic radiology facilities to determine the radiation levels in and around work areas and radiology equipment, and to assess the radiation protection of the workplace and individuals (see Chapter 24). Such monitoring devices should typically measure in integrate mode and for direct (or real time) measurements include ionization chambers and some specifically developed semiconductor detectors suitable for scattered radiation. Longer term monitoring is typically achieved with film, or increasingly with solid state devices (for personal dosimeters (such as TLDs or film badges), see Section 21.5).

The use of survey meters (such as Geiger Müller counters or proportional counters) is not recommended for diagnostic radiology. Such devices are typically designed to detect isotope emissions; they are used extensively for radiation detection in nuclear medicine and have some application in radiation therapy, particularly for ^{60}Co units, brachytherapy usage and radioactive iodine treatment of patients. The two main difficulties in diagnostic radiology for the use of Geiger Müller counters, for example, is the response time of several seconds, when diagnostic X ray exposures have a duration of only small fractions of a second, and they also exhibit a strong energy dependence at low photon energies. Detailed descriptions of these instruments can be found elsewhere [21.5].

Detectors used for occupational and public exposure measurement should be traceable to appropriate calibration standards at suitable X ray energies (e.g. ISO Narrow series N40 to N80 [21.6]). While the user should know or estimate the mean energy of the X ray beam for calibration factor application, in some situations, such as the mean energy of radiation transmitted through protective barriers, this can be difficult. In these cases, it is acceptable to use measurements directly from a detector with a small energy response variation. The uncertainty of measurement should be assessed with reference to the variation in the calibration coefficients within the energy range used.

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